(19) Europäisches Patentamt
European Patent Office
Office européen des brevets

(11) EP 0 896 228 A1

(12) EUROPEAN PATENT APPLICATION

(43) Date of publication:
10.02.1999 Bulletin 1999/06

(51) Int Cl.6: G01R 33/421

(21) Application number: 98305326.5

(22) Date of filing: 03.07.1998

(84) Designated Contracting States:
AT BE CH CY DE DK ES FI FR GB GR IE IT LI LU MC NL PT SE
Designated Extension States:
AL LT LV MK RO SI

(30) Priority: 07.08.1997 US 908359

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(54) Self-shielded gradient coil assembly and method of manufacturing the same

(57) A superconducting magnetic imaging apparatus includes a vacuum vessel **(16)** having a central helium reservoir **(20)** in which superconducting magnetic coil windings **(12)** are maintained at a superconducting temperature. The vacuum vessel defines an internal bore **(22)** within which a self-shielded gradient coil assembly **(26)** and an RF coil **(30)** are received. The self-shielded coil assembly includes a single former **(50)** which defines an imaging region **(14)** within which an imaged portion of a subject is received. Primary x, y, and z-gradient coils **(72-76)** are positioned over an RF ground screen **(70)** that is bonded to the former **(50)**. A number of comb-like spacers **(84)** extend from the former to supporting shield x, y, and z-gradient coils **(110-114)**. The comb-like spacers define passages between the primary and secondary gradient coils which receive inner and outer cooling tubes **(90, 92)** and shim tray molds **(108)**. The fully assembled gradient coil assembly is potted to form a unitary structure in a single potting step. The shim tray molds are subsequently removed to define channels or pockets suitable for receiving shim trays **(118)**.

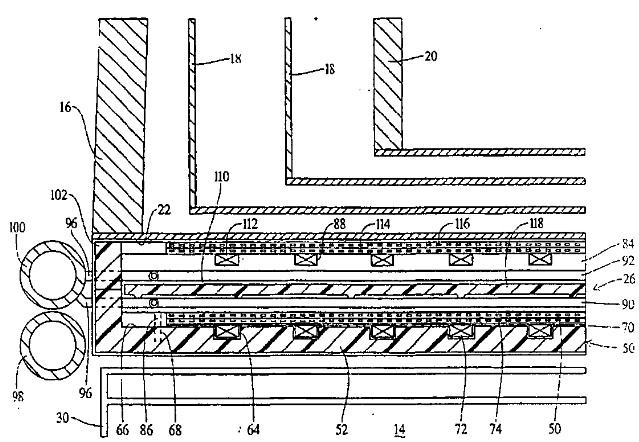

Fig.2

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Description

[0001] The present invention relates to magnetic resonance imaging. It finds particular application in conjunction with gradient coils for magnetic resonance imaging apparatus and will be described with particular reference thereto. However, it should be appreciated that the present invention may also find application in conjunction with magnetic resonance spectroscopy systems and other applications which utilize gradient magnetic fields.

[0002] In magnetic resonance imaging (MRI) applications, three orthogonal gradient fields are employed to provide spatial resolution by frequency discrimination of an MRI signal. A gradient coil set typically includes three discrete gradient coils for generating the x, y, and z-gradient fields. The coils are insulated from each other and are layered on a cylindrical former. Commonly, the entire gradient coil set is overwrapped and epoxy impregnated for greater structural strength to withstand the warping forces when the current carrying conductors of the coils interact with the primary magnetic field of an MRI apparatus.

[0003] The gradient coils are commonly pulsed with current pulses having a short rise time and a high duty cycle. Pulsing the gradient coils produces magnetic field gradients across the imaging region. as well as magnetic field gradients which interact with external metallic structures such as cold shields in a superconducting magnet. This intcraction generates eddy currents in the cold shields. which. in turn. generate eddy magnetic fields. The eddy fields have a deleterious effect on the temporal and spatial quality of the magnetic field in the examination region, hence in the resultant image quality.

[0004] One approach for circumventing the eddy current problem is to place a secondary or shielding gradient coil set between the primary gradient coil set and the cold shields. The shielding gradient coil set substantially zeroes the magnetic field externally thereof, thus preventing the formation of eddy currents in the cold shields.

[0005] A unitary, self-shielded gradient coil assembly typically includes a secondary or shielding gradient coil set spaced radially outwardly of a primary gradient coil set and driven in series therewith. The primary and secondary gradient coil sets include insulated coils for generating x, y, and z-gradient fields which are layered on separate cylindrical formers. The primary and secondary gradient coil sets are individually overwrapped and epoxy impregnated for greater structural strength. A mechanical means typically connects the two formers together to form a unitary structure while maintaining the coil sets in a spaced relationship.

[0006] Typically, main superconducting magnets have bore diameters of least 90cm and are at least 1.8m in length. Magnets of such size can accommodate shielded gradient coil structures that are constructed from two separate formers (wherein each former has a thickness of approximately 5.0 to 7.5 cm) because there is adequate space between the predetermined patient aperture and the inside bore diameter of the magnet to accommodate such a radial build.

[0007] Reducing the diameter of the magnet bore dramatically reduces magnet costs. However, reducing the patient receiving bore reduces the utility of the MRI apparatus and its commercial acceptability.

[0008] In addition, known methods of manufacturing self-shielded gradient coil assemblies have a time-consuming, final alignment step. Alignment tooling and fixtures are used to ensure that the axial isocentre of the primary gradient coils coincide with the axial isocentre of the secondary gradient coils.

[0009] The present invention contemplates a new self-shielded gradient coil assembly and method for manufacturing such a self-shielded gradient coil assembly which overcomes the above-referenced problems and others.

[0010] In accordance with one aspect of the present invention, there is provided a magnetic resonance imaging apparatus. The MRI apparatus includes a main field magnet for generating a temporally constant magnetic field through an examination region. A radio frequency transmitter excites and manipulates magnetic resonance in selected dipoles in the examination region. A receiver demodulates magnetic resonance signals received from the examination region. A processor reconstructs the demodulated resonance signals into an image representation. A self-shielded gradient coil assembly induces gradient magnetic fields across the temporally constant magnetic field. Characterized in that the self-shielded gradient coil assembly includes a former, primary gradient coils supported by the former, and a number of combs extending from the former for supporting shielding gradient coils.

[0011] In accordance with a second aspect of the present invention, there is provided a method of manufacturing a self-shielded gradient coil assembly for use in a magnetic resonance imaging apparatus including a main field magnet for generating a temporally constant magnetic field through an examination region. A radio frequency transmitter excites and manipulates magnetic resonance in selected dipoles in the examination region. A receiver demodulates magnetic resonance signals received from the examination region. A processor reconstructs the demodulated resonance signals into an image representation. The self-shielded gradient coil assembly induces gradient magnetic fields across the temporally constant magnetic field. The method includes joining a number of primary gradient coils to a cylindrical former, joining a number of combs to the former, joining a number of shielding gradient coils to the combs to form a unitary structure, and potting the unitary structure.

[0012] The invention will now be described in detail, by way of example, with reference to the accompanying

drawings in which:

Figure 1 is a diagrammatic illustration of a magnetic resonance imaging system having a self-shielded gradient coil assembly in accordance with a first embodiment of the present invention;

Figure 2 is an enlarged fragmentary, sectional view of the self-shielded gradient coil assembly of Figure 1;

Figure 3 is a perspective view of the self-shielded gradient coil assembly of Figure 2, partially assembled with a primary z-gradient coil wound over an RF ground screen that is secured to a former;

Figure 4 is a perspective view of the self-shielded gradient coil assembly of Figure 3 with a primary x-gradient coil secured over the z-gradient coil, and a primary y-gradient coil positioned over the x-gradient coil;

Figure 5 is a perspective view of the self-shielded gradient coil assembly of Figure 4 with a plurality of combs secured over the y-gradient coil, and a ring secured over a service-end flange;

Figure 6 is a perspective view of the self-shielded gradient coil assembly of Figure 5 with inner and outer cooling tubes positioned between adjacent combs, and service-end coolant manifolds joined to free ends of the cooling tubes:

Figure 7 is a perspective view of the self-shielded gradient coil assembly of Figure 6 with a plurality of shim tray pockets positioned between the inner and outer cooling tubes, and a secondary z-gradient coil wound over the combs:

Figure 8 is a perspective view of the self-shielded gradient coil assembly of Figure 7 with a secondary x-gradient coil secured over the secondary z-gradient coil, and a secondary y-gradient coil secured over the secondary x-gradient coil;

Figure 9 is a perspective view of the fully- assembled and potted self-shielded gradient coil assembly in accordance with the first embodiment of the present invention; and

Figure 10 is a perspective view of a partially assembled, self-shielded gradient coil assembly in accordance with a second embodiment of the present invention, wherein a primary z-gradient coil is wound over an RF ground screen that is temporarily secured to a removable mandrel.

[0013] With reference to Figure 1, a main magnetic field control **10** controls superconducting or resistive magnets **12** such that a substantially uniform, temporally constant magnetic field is created along a z-axis through an examination region **14**. Preferably, a vacuum vessel **16** surrounds one or more cold shields **18** which surround a central helium reservoir **20** in which superconducting magnets **12** are maintained at a superconducting temperature. The vacuum vessel defines an internal bore **22**.

[0014] A magnetic resonance echo means applies a series of radio frequency (RF) and magnetic field gradient pulses to invert or excite magnetic spins, induce magnetic resonance, refocus magnetic resonance. manipulate magnetic resonance, spatially and otherwise encode the magnetic resonance, to saturate spins, and the like to generate magnetic resonance imaging and spectroscopy sequences.

[0015] More specifically, gradient pulse amplifiers **24** apply current pulses to selected pairs of whole body gradient coils of a self-shielded gradient coil assembly **26** to create magnetic field gradients along x. y, and z-axes of the examination region **14**. The gradient coil assembly **26** is positioned within the internal bore **22**. A digital radio frequency transmitter **28** transmits radio frequency pulses to a whole body RF coil **30** to transmit RF pulses into the examination region. The RF pulses are used to saturate, excite resonance, invert magnetization, refocus resonance. or manipulate resonance in selected portions of the examination region. A scan or sequence controller **32** controls the gradient amplifiers and the digital receiver amplifies an imaging sequence, e.g. a fast spin echo sequence in which each resonance excitation or shot is followed by a plurality or train of echoes.

[0016] For whole body applications, the resonance signals are commonly picked up by the whole body RF coil **30** and demodulated by a receiver **34** to form data lines that are stored in a data line memory **36**.

[0017] An image reconstruction processor **38** reconstructs the data set into an image representation that is stored in an image memory **40**. A video processor **42** converts selected data from the image memory into an appropriate format for display as slices. three dimensional rendering and the like on a human-readable monitor **44**.

[0018] Referring now to Figures 2 and 3, the self-shielded gradient coil assembly **26** includes an electrically insulating, hollow, right cylinder coil-form or former **50**. In the embodiment being described, the former **50** includes is a glass filament wound resin tube (FWT) **52** and a first flange **54** extending transversely from a front or patient-end of the cylindrical portion **52**. The first flange **54** includes a plurality of circumferentially spaced apertures **56** extending therethrough.

[0019] The former **50** also includes a second flange **58** extending transversely at a rear or service-end of the cylindrical portion **52**. The second flange **58** includes a plurality of paired notches **60** circumferentially spaced along its outer surface. As best seen in Figure 2, a plu-

rality of axially spaced grooves **64** extend circumferentially around an outer surface **66** of the cylindrical portion **52**. In addition, a plurality of radially-oriented, closed-end bores **68** (one shown) extend into the cylindrical portion **52** from the outer surface thereof.

[0020] In the embodiment being described, an integral RF ground screen **70** is bonded to the outer surface of the cylindrical portion **52** and urged into the axially-spaced grooves **64**, as shown in Figure 2. The RF ground screen **70** is preferably bonded to the former using a wet laying or a vacuum pressure impregnation process (VPI). The RF ground screen is preferably formed from phosphor and bronze materials which are matched to the slew rate design, as is known in the art. Alternatively, the RF ground screen can be mounted to the inner diameter of the former **50**. or embedded with the filament windings in the tube defining the former.

[0021] A primary z-gradient coil **72** is wound into the grooves **64** over the RF screen **70**. Optionally, the z-gradient coil is temporarily or spot bonded to the cylindrical portion **52**. Alternatively, the z-gradient coil can be wound into the grooves **64** and the RF screen **70** can be bonded to the cylindrical portion **52** over the z-gradient coil. The embodiment being described incorporates a bunched primary z-gradient coil design. However, a distributed current coil design can also be utilized.

[0022] With continuing reference to Figure 2 and particular reference to Figure 4, a primary x-gradient coil assembly **74** is temporarily spot bonded to the gradient coil assembly **26** over the primary z-gradient coil **72**. A primary y-gradient coil **76** is then temporarily spot bonded to the gradient coil assembly **26** over the primary x-gradient coil assembly **74**. In the embodiment being described, the x- and y-gradient coil assemblies **74, 76** each include four (4) electric conductor distributed "thumbprint" quadrant windings (not shown) mounted in pairs on flexible dielectric sheets. For both gradient coils **74, 76,** two of the quadrant windings are adjacent and extend around a first portion of the former, while the other two quadrant windings are adjacent and extend around a second portion of the former.

[0023] Thus, the x- and y-gradient coils **74, 76** are substantially equivalent to each other, but one of the gradient coils **74, 76** is circumferentially offset from the other gradient coil by approximately 90° relative to the central axis of the former. The quadrant winding sections of the y-gradient coils. are mounted around the x-gradient coil **74**. The opposing edges **78** and opposing edges **80** (one shown) are positioned adjacent to each other around the former, such that the edges **78, 80** lie in a vertical (y, z) plane.

[0024] The quadrant winding sections of the x-gradient coil **74** are mounted around the z-gradient coil **72** with opposing edges **82** positioned adjacent to each other. The x-gradient coil **74** is circumferentially offset from the y-gradient coil **76** by 90°. For the x-gradient coil, the edges **82** and the opposite edges (not shown) lie in a horizontal (x, z) plane.

[0025] With continuing reference to Figure 2, and particular reference to Figure 5, a plurality of circumferentially spaced-apart dielectric combs **84** are secured to the gradient coil assembly **26** over the y-gradient coil **76**. Each of the combs **84** includes one or more pins **86** (Fig. 2) extending from a lower surface thereof into the apertures **68** of the former **50**. Other mechanical mounting arrangements, such as slots in the end flanges, are also contemplated. The combs **84** also include a plurality of axially spaced notches **88** along their outer surfaces. In the preferred embodiment, the combs **84** are formed from a glass-resin composite material.

[0026] The combs **84** support the rest of the radial build-up of the gradient coil assembly **26** as described further below. Thus, it should be appreciated that the combs **84** replace the second filament wound tube or former associated with prior self-shielded gradient coil assemblies. In order to avoid aligning the primary and secondary (described below) gradient coil sets. the combs **84** (and hence the notches **88**). as well as the tooling for forming the grooves **64**. is located or positioned precisely relative to the machined surface of the flange **54**. Thus. both primary and secondary coil sets are mechanically aligned within the machine tool tolerances.

[0027] With continuing reference to Figure 2, and particular reference to Figure 6, an inner serpentine, e.g. M-shaped. cooling tube or heat exchanger **90** is spaced radially inward from an outer serpentine, e.g. M-shaped cooling tube or heat exchanger **92** between adjacent combs **84**. Each of the cooling tubes **90, 92** is temporarily spot bonded to the gradient coil assembly **26**.

[0028] Each of the cooling tubes **90, 92** includes an inlet **94** and an outlet **96**. Each of the inlets **94** is joined to a single, circular, service-end coolant supply manifold **98**; each of the outlets **96** is joined to a single, circular service-end coolant return manifold **100**. In the embodiment being described, the cooling tubes **90, 92,** and manifolds **98, 100** are seamless and are formed from stainless steel for heat strength and minimal magnetic susceptibility interference during imaging.

[0029] The inlets **94** and outlets **96** pass through the paired notches **60** on the service-end flange **58**. After the cooling tubes are in place, a service-end ring **102** is then bonded over the flange **58** such that notches **104** associated with the service-end ring **102** align with the cooling tube inlets and outlets **94, 96**. Further. when the service-end ring **102** is joined to the flange **58,** apertures **106** are formed between adjacent pairs of notches **60** and **104**.

[0030] As described below. shim tray molds extend through the apertures **56** to form channels or pockets which extend axially through the gradient coil assembly **26** from the service-end to the patient receiving-end. The service-end ring **102** may be formed from one or more arcuate members which, when joined to the flange **58,** form a substantially continuous circumference.

[0031] With continuing reference to Figure 2, and par-

ticular reference to Figure 7, a plurality of shim tray molds **108** are inserted through the apertures **56** of the flange **54** into the gaps between the inner and outer cooling tubes **90, 92** and through the apertures **106**. Prior to insertion into the gaps, each mold **108** is wrapped with approximately three sheets of woven glass material to provide structural integrity around the resultant channel or pocket **110** (Fig. 2) that is formed when the molds **108** are subsequently removed following the single potting step described below.

[0032] The shim tray molds **108** may be solid and formed from a non-ferrous material such as aluminum, wax, etc. so as not to leave a ferrous residue when removed. The shim tray molds **108** have a highly polished surface with a taper of approximately 2mm from end to end to facilitate removal through apertures **56** with a hydraulic removal apparatus.

[0033] Secondary or shielding z-gradient coil windings **112** are then wound into the grooves **88** of the combs **84**. The embodiment being described incorporates a distributed z-gradient shielding coil design. As shown in Figures 2 and 8, a secondary or shielding x-gradient coil **114** and a secondary or shielding y-gradient coil **116** are temporarily spot bonded to the gradient coil assembly **26** over the windings **112**. The secondary shielding coils **112, 114** are structurally similar to. and oriented in the same manner as the primary gradient coils **74, 76,** as described above.

[0034] If the primary and secondary gradient coils are to be driven in series, the electrical interconnections are then made. A potting mold (not shown) is placed around the gradient coil assembly **26**. A potting compound. such as epoxy, is drawn into and fills all voids within the gradient coil assembly in a single potting step. After the epoxy is cured and the potting mold removed, the shim tray molds **108** are removed through the apertures **56**. The resulting cavities form the continuous channels or pockets **110** which extend from the service-end of the gradient coil assembly to the patient-end thereof. The channels **110** receive shim trays **118** which correct for limited non-uniformities of the primary magnetic field and render the primary magnetic field substantially uniform within the bore. The channels also provide air flow paths to promote cooling of the shim trays to minimize shim steel susceptibility changes.

[0035] Referring now to Figure 10, a self-shielded gradient coil assembly **120** is manufactured without incorporating any filament wound tubes or formers. In particular, a first flange **122**, a second flange **124**, and an RF ground screen **126** are initially positioned over a removable mandrel **128**. The removable mandrel with a highly polished surface has a slight taper from end to end to facilitate the subsequent removal after the final potting step is completed. A primary z-gradient coil **130** is then wound on the RF ground screen **126**. The radial build-up of the gradient coil assembly **120** then continues in the same manner as described above in conjunction with Figures 4-9.

[0036] Thus, the self-shielded gradient coil assembly incorporates primary and shielding coils which are aligned and potted into a single, unitary construction. Because the primary and secondary assemblies are a unitary construction the axial alignment of the primary and secondary coils is built in within the mechanical tolerances of the machining equipment as the assembly is built. There is no need to perform an alignment procedure between separate primary and secondary gradient coil constructions. Thus, the prior art alignment process is eliminated: and the time required to perform the alignment is saved.

[0037] It should be appreciated, that the order of the transverse (e.g., x- and y-gradient coils) and the axial (e.g., z-gradient coils may be varied or reversed. In addition, the inner cooling tubes **90** may be positioned between the former and the primary gradient coils, or positioned in any gap between the transverse and axial primary gradient coils. Likewise, the outer cooling tubes **92** may be positioned in any gap between the transverse and axial secondary gradient coils.

[0038] Further, the position of the combs **84**, and the tooling for forming the notches **64**, may be located from a surface other than the patient-end flange **54**. Different bonding methods may be employed to secure the respective components to the gradient coil assembly. For instance. a heat cured epoxy and/or an epoxy which uses a catalytic agent, may be used.

[0039] One advantage of the illustrated embodiment is that the number of formers in a self-shielded gradient coil assembly is reduced. Another advantage resides in reduced cost. Another advantage is that the step of aligning a primary gradient coil set with a secondary gradient coil set is eliminated. Another advantage is that stored magnetic energy is minimised consistent with gradient strength and uniformity requirements. Further advantages reside in improved product quality, gradient strength, and slew rates.

Claims

1. Magnetic resonance imaging apparatus including a main field magnet **(12)** for generating a temporally constant magnetic field through an examination region **(14)**, a radio frequency transmitter **(28)** for exciting and manipulating magnetic resonance in selected dipoles in the examination region, a receiver **(34)** for demodulating magnetic resonance signals received from the examination region, a processor **(38)** for reconstructing the demodulated resonance signals into an image representation, and a self-shielded gradient coil assembly **(26)** for inducing gradient magnetic fields across the temporally constant magnetic field, the self-shielded gradient coil assembly **(26)** further including: a former **(50)**; primary gradient coils **(72-76)** supported by said former; and a plurality of combs **(84)** extending from

said former for supporting shielding gradient coils **(112-114)**.

2. Magnetic resonance imaging apparatus as claimed in claim 1, in which the shielding gradient coils **(112-114)** are radially displaced from the primary gradient coils **(72-74)** by the combs **(84)** to define a passage therebetween.

3. Magnetic resonance imaging apparatus as claimed in claim 2, in which the self-shielded gradient coil assembly **(26)** further includes a plurality of cooling tubes **(90, 92)** potted in said passage.

4. Magnetic resonance imaging apparatus as claimed in claim 2 or claim 3, in which the self-shielded gradient coil assembly **(26)** further includes a plurality of stainless steel cooling tubes **(90, 92)** disposed in the passage, the cooling tubes having an inlet **(94)** connected to a supply manifold **(98)** and an outlet **(96)** connected to a return manifold **(100)**.

5. Magnetic resonance imaging apparatus as claimed in any one of claims 2 to 4, in which the self-shielded gradient coil assembly **(26)** further includes a plurality of channels **(110)** defined within said passage; and a plurality of shim trays **(118)** disposed in the plurality of channels.

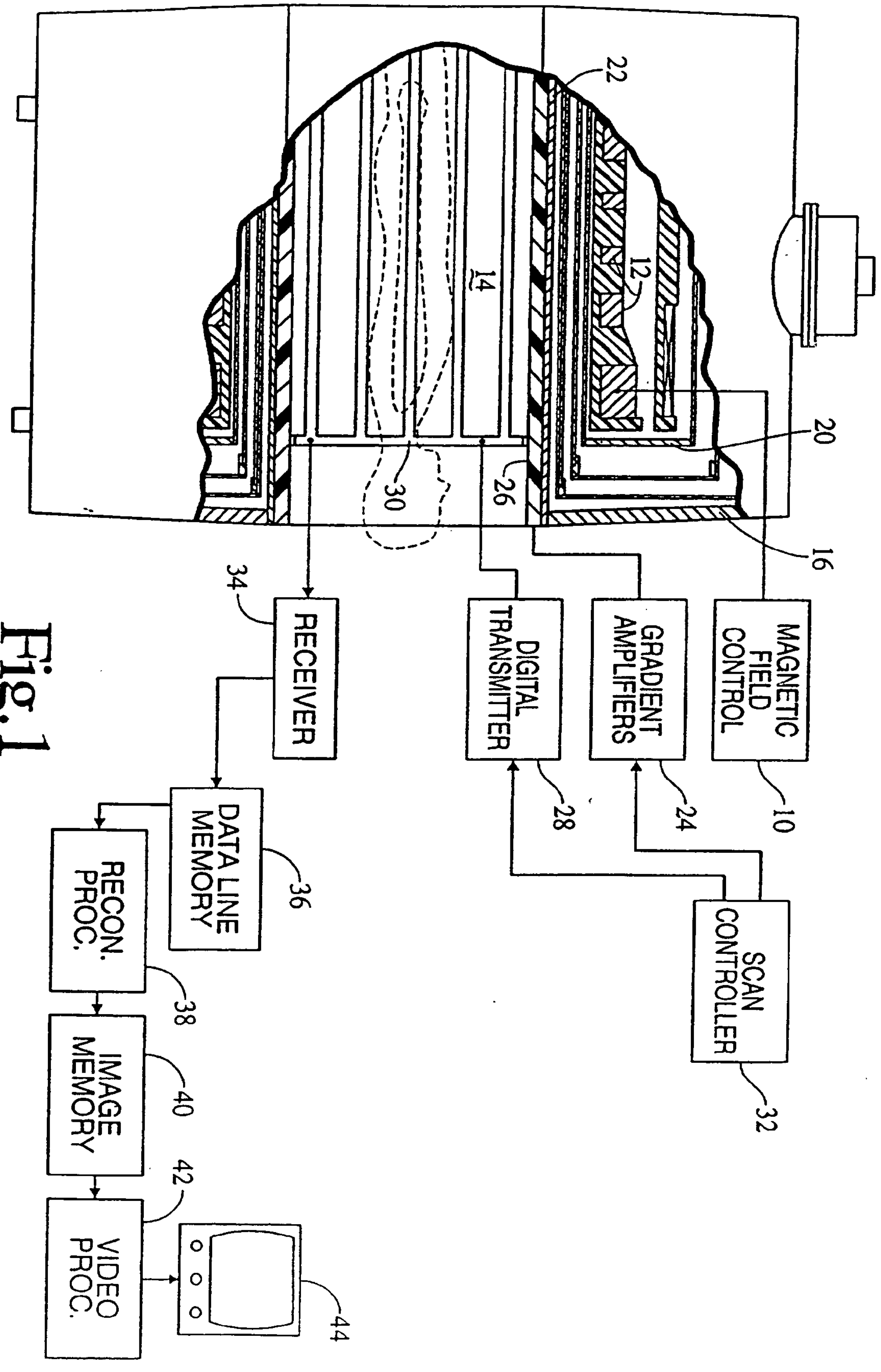
6. Magnetic resonance imaging apparatus as claimed in any one of claims 2 to 5, in which the self-shielded gradient coil assembly **(26)** further includes: a plurality of inner cooling tubes **(90)** positioned within said passage adjacent the primary gradient coils **(72-76)**; a plurality of outer cooling tubes **(92)** positioned within said passage adjacent the secondary gradient coils **(112-114)**; and a plurality of channels **(110)** interposed between said inner and outer cooling tubes.

7. Magnetic resonance imaging apparatus as claimed in any one of claims 1 to 6, in which the former **(50)** includes grooves **(64)** therein for receiving a primary z-gradient coil **(72)**, and said combs **(84)** each include notches **(88)** for receiving a shielding z-gradient coil **(112)** therein.

8. Magnetic resonance imaging apparatus as claimed in any one of claims 1 to 7, in which the combs **(84)** and the former **(50)** are connected by a pin assembly **(68, 86)** which positions notches **(88)** in the combs for receiving the shield coil windings **(112-114)** relative to the former.

9. A method of manufacturing a self-shielded gradient coil assembly **(26)** for use in a magnetic resonance imaging apparatus including a main field magnet **(12)** for generating a temporally constant magnetic field through an examination region **(14)**, a radio frequency transmitter **(28)** for exciting and manipulating magnetic resonance in selected dipoles in the examination region, a receiver **(34)** for demodulating magnetic resonance signals received from the examination region, a processor **(38)** for reconstructing the demodulated resonance signals into an image representation, the self-shielded gradient coil assembly **(26)** inducing gradient magnetic fields across the temporally constant magnetic field, the method comprising: supporting a plurality of primary gradient coils **(72-74)** to a cylindrical former **(50)**; supporting a plurality of comb-like spacers **(84)** on the former; supporting a plurality of shield gradient coils **(112-114)** to the spacers; and potting the former, the primary gradient coils, the spacers and the shield gradient coils in resin to form a unitary construction.

10. A method of manufacturing a self-shielded gradient coil assembly as claimed in claim 9. further comprising: positioning the unitary structure within a bore of the main field magnet **(12)** defining the examination region **(14)**.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DATA LINE MEMORY
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RECON. PROC.
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IMAGE MEMORY
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VIDEO PROC.
44

Fig.1

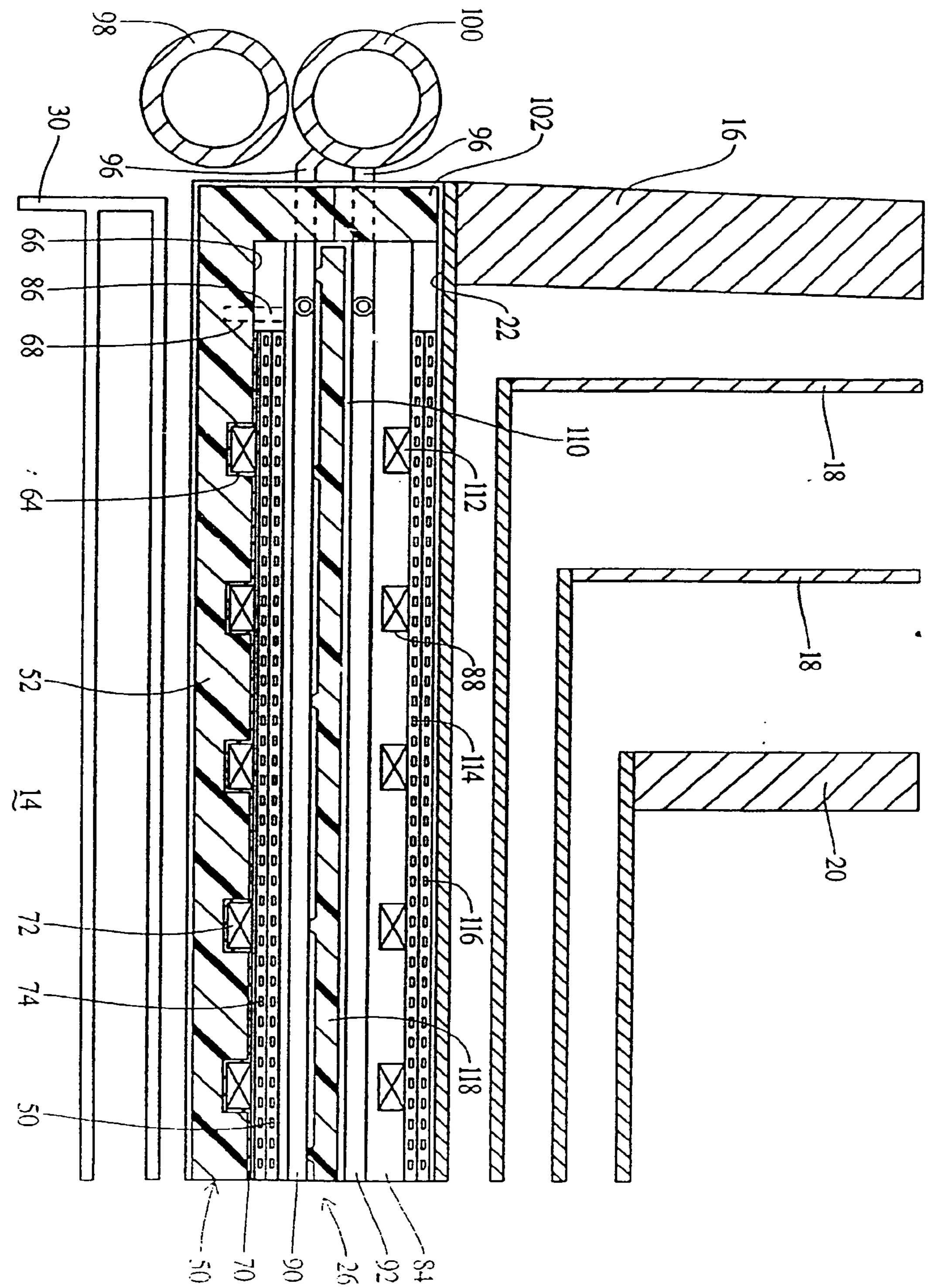
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Fig.2

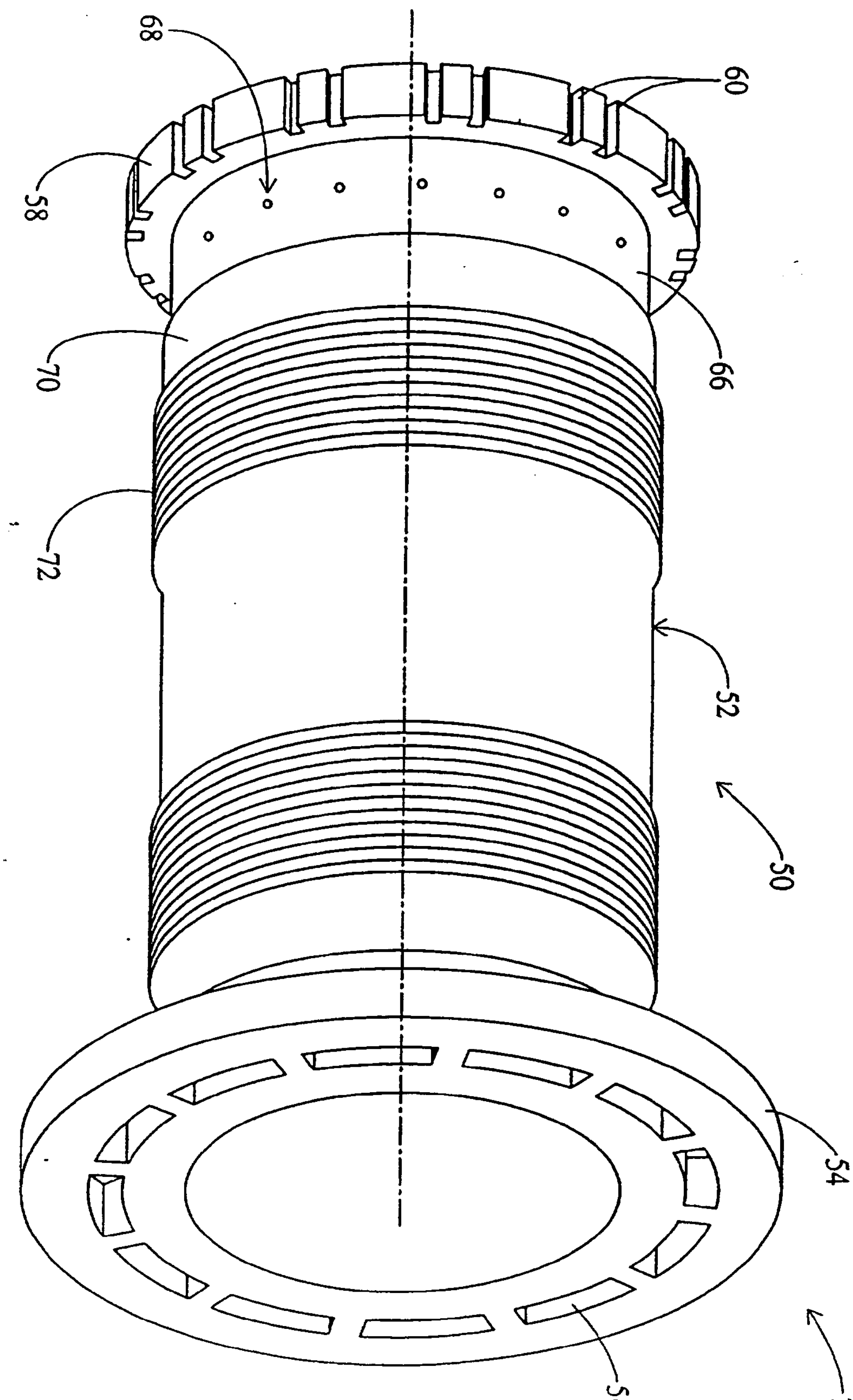

Fig. 3

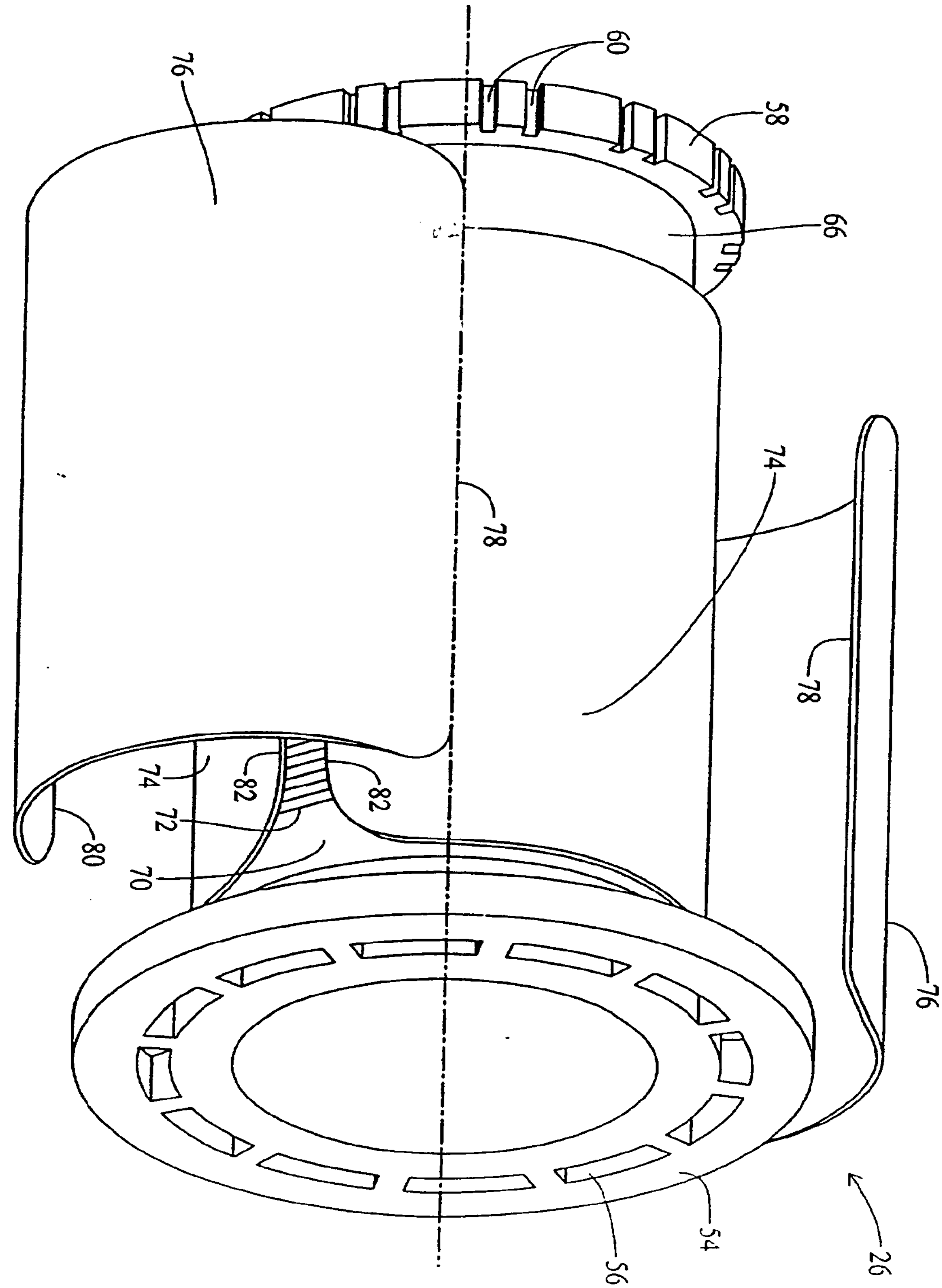

Fig. 4

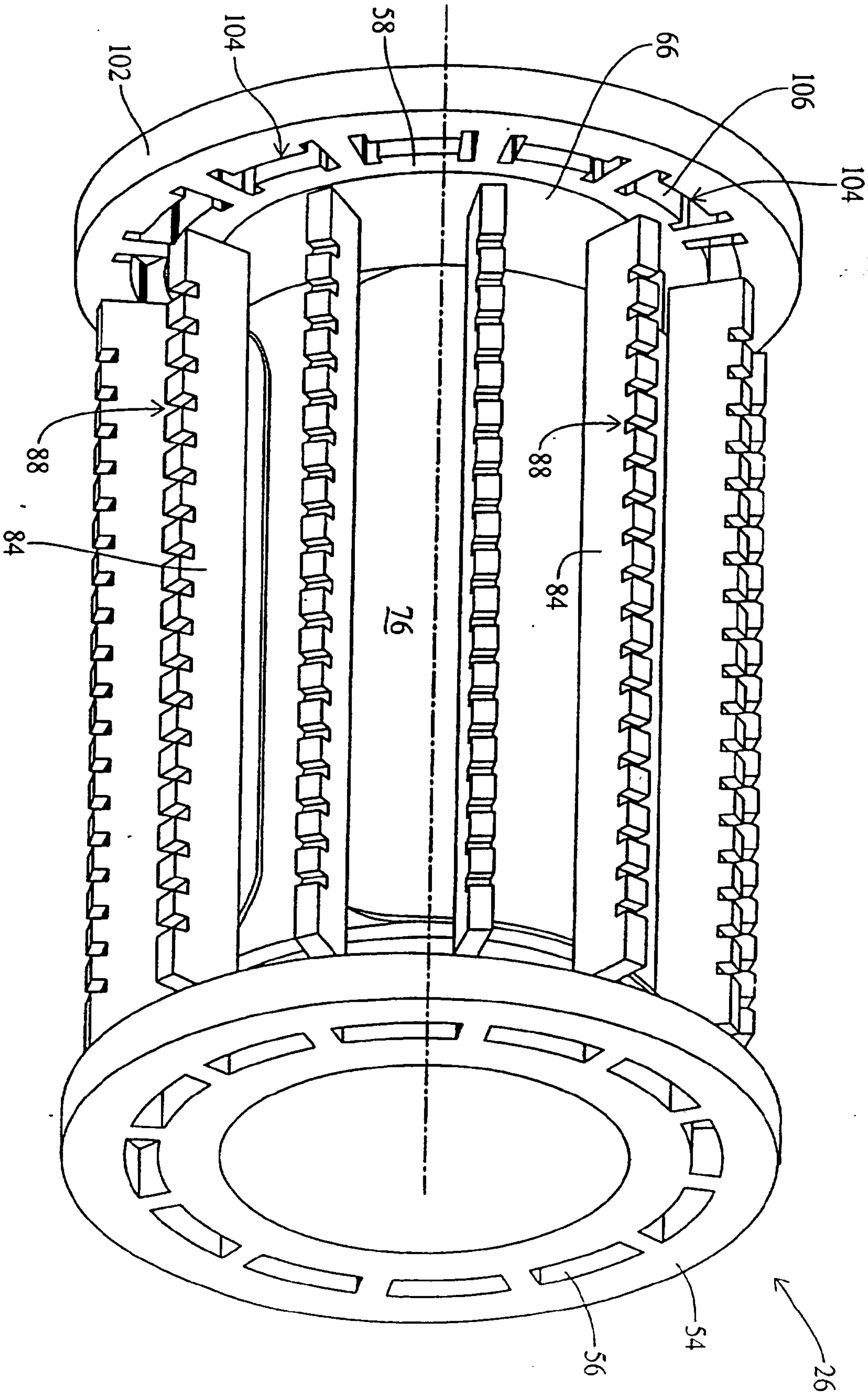

Fig. 5

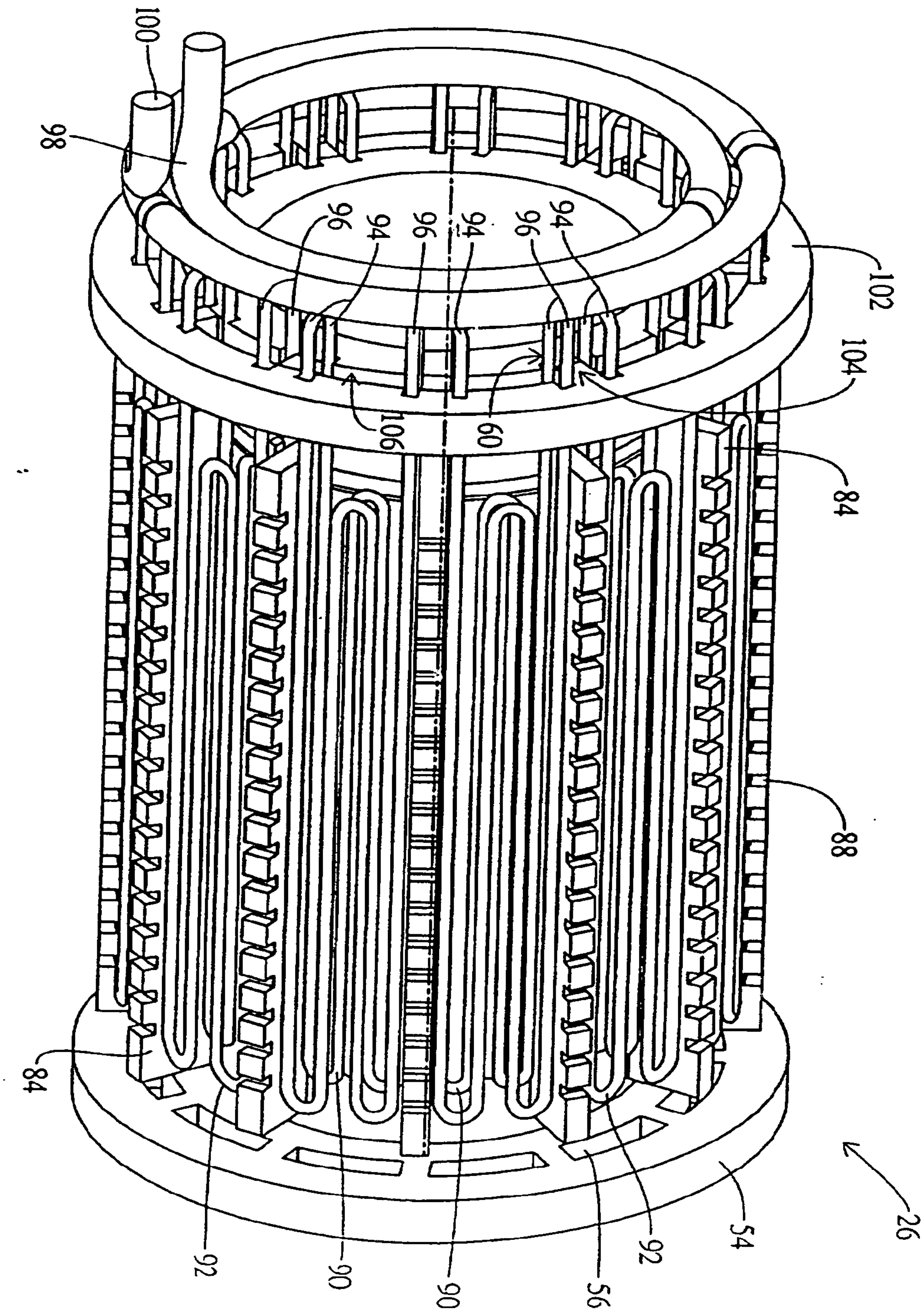

Fig.6

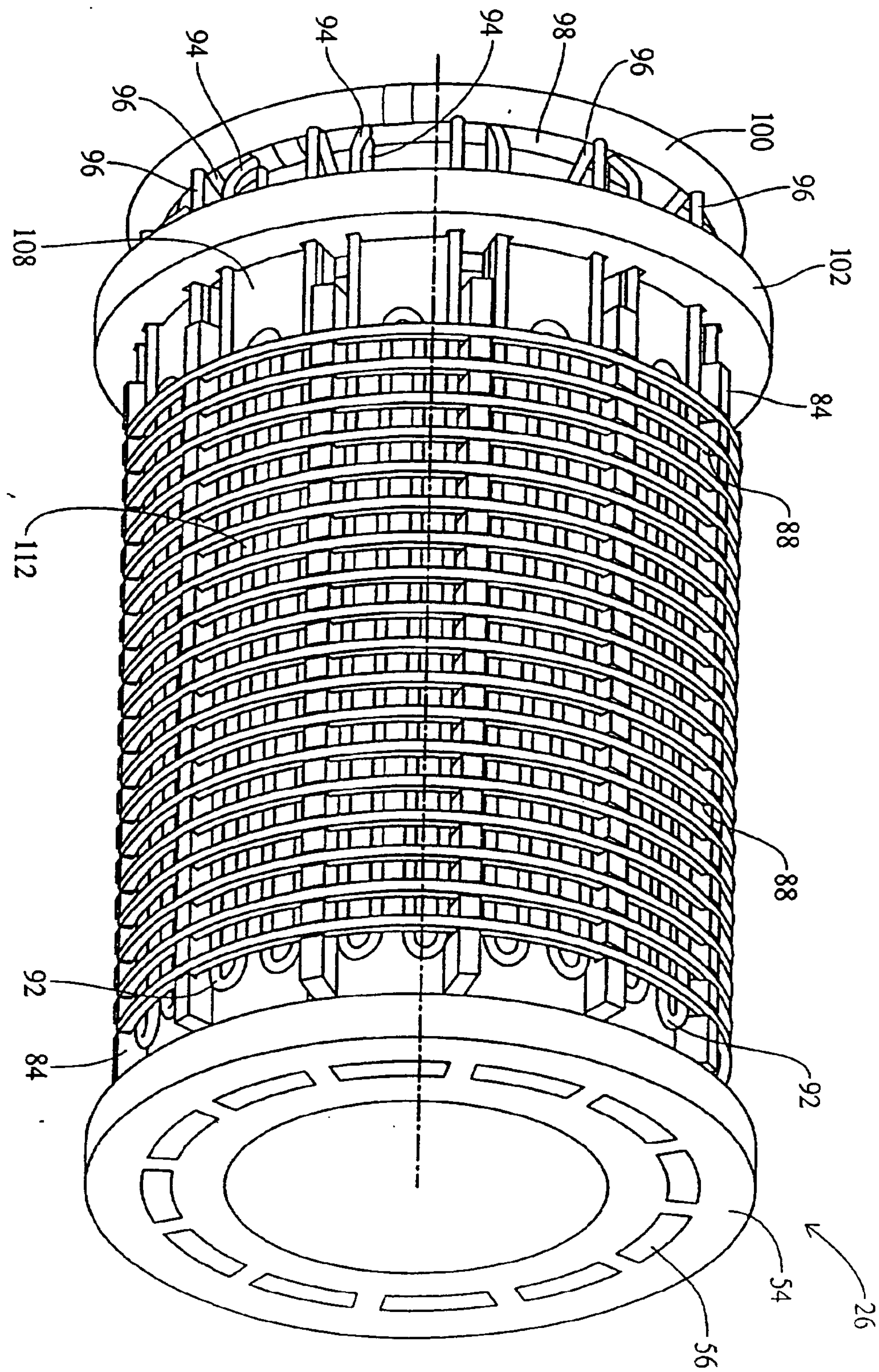

Fig.7

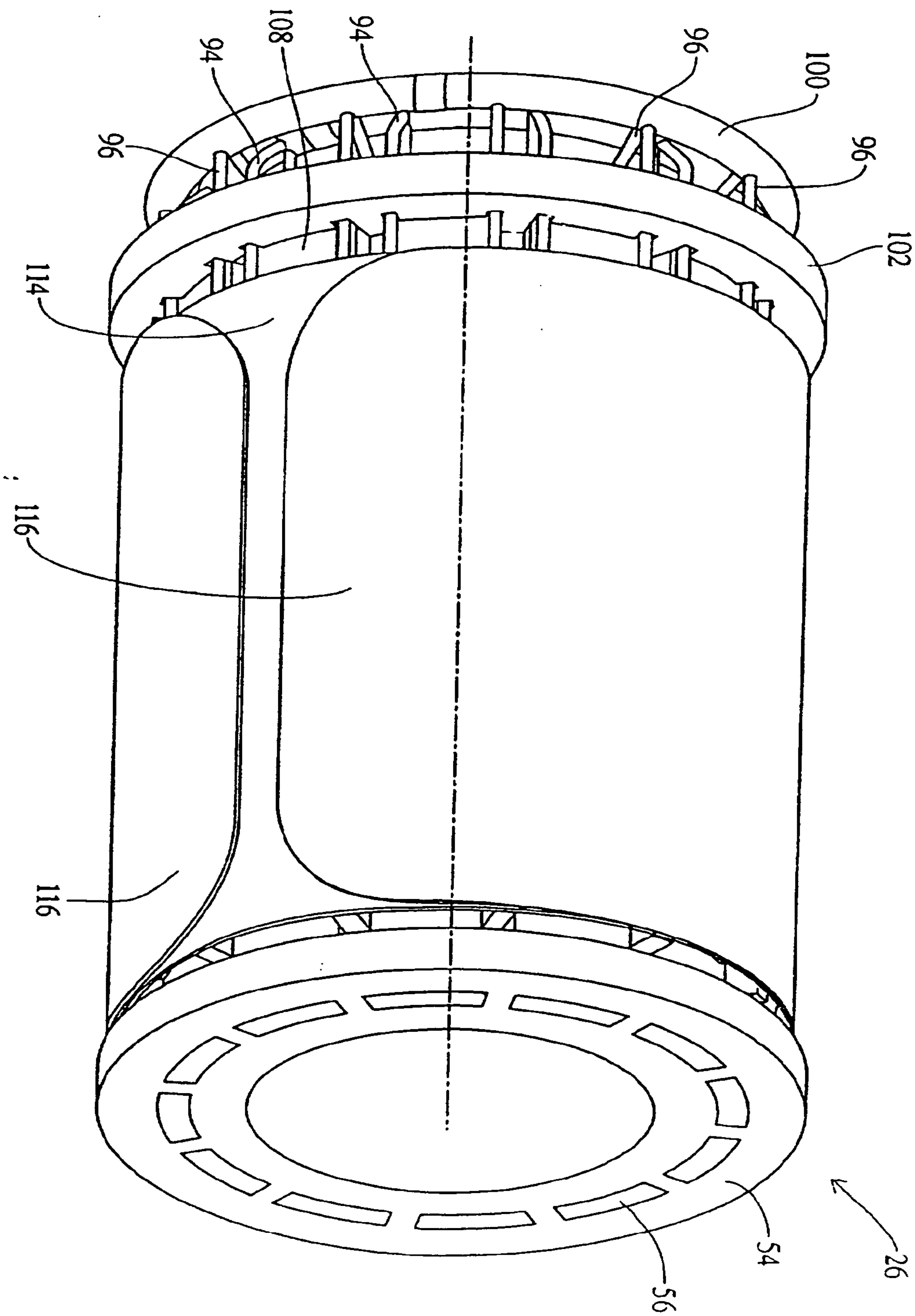

Fig.8

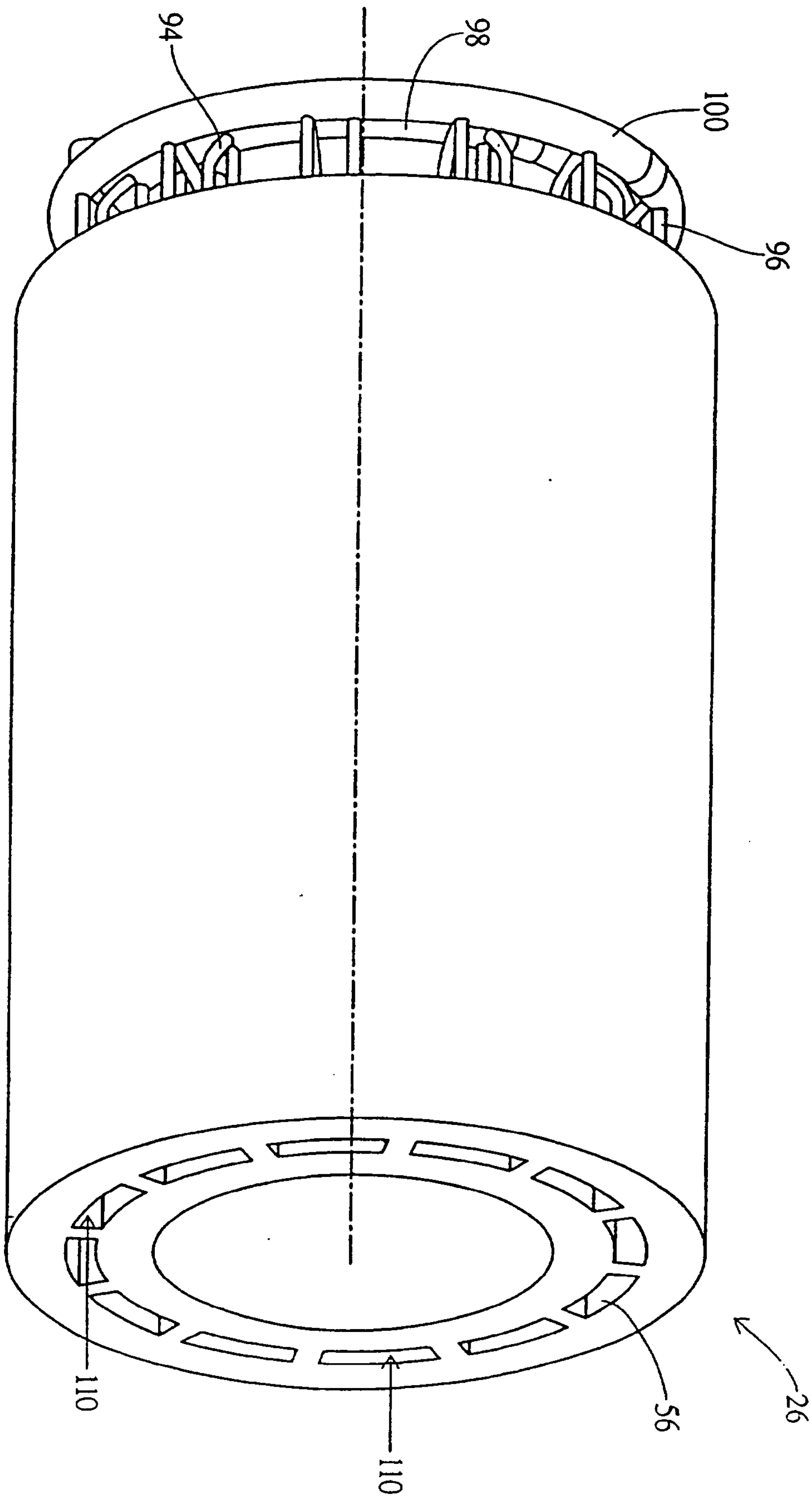

Fig. 9

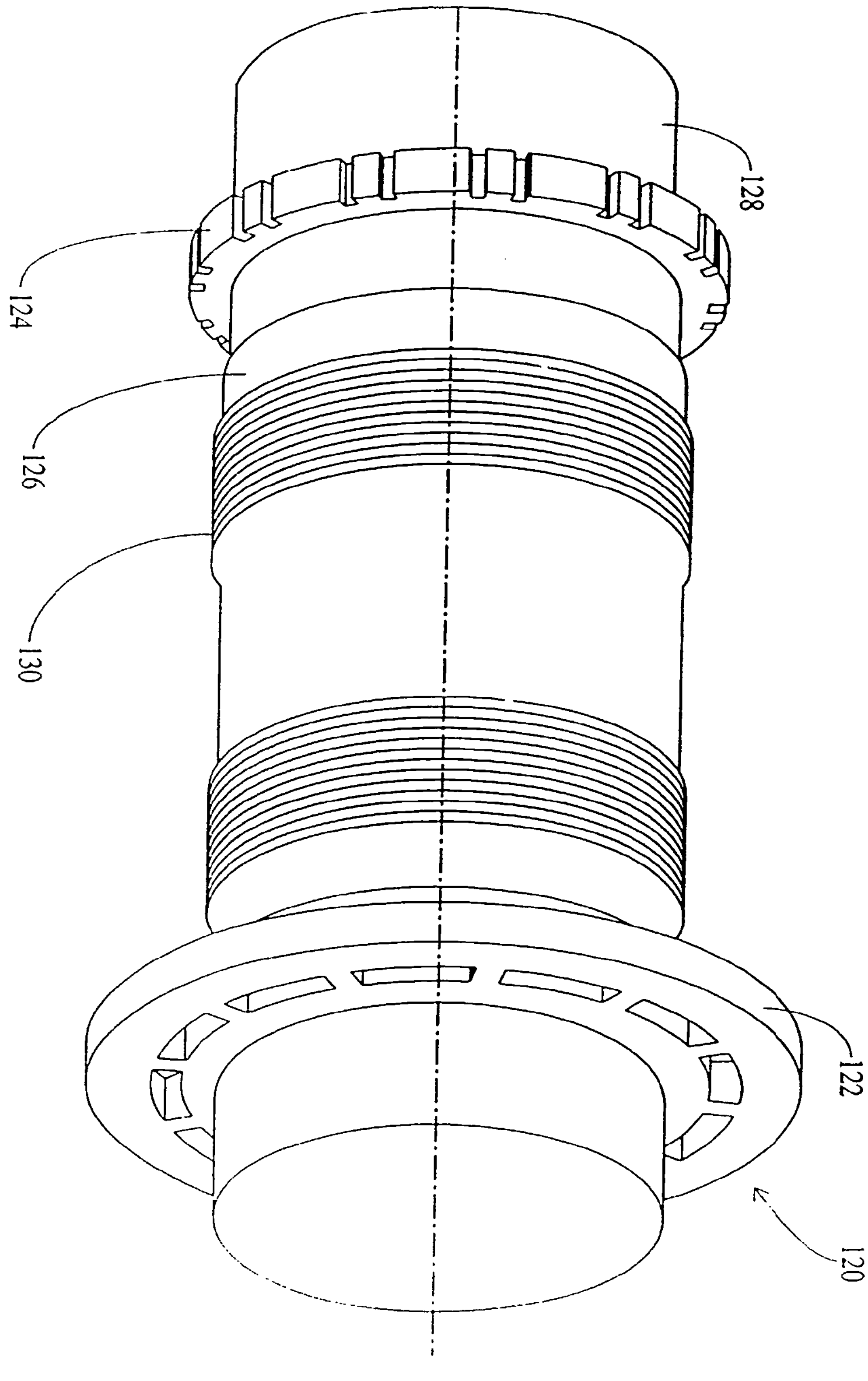

Fig.10

European Patent Office

EUROPEAN SEARCH REPORT

Application Number
EP 98 30 5326

DOCUMENTS CONSIDERED TO BE RELEVANT

Category	Citation of document with indication, where appropriate, of relevant passages	Relevant to claim	CLASSIFICATION OF THE APPLICATION (Int.Cl.6)
X	US 5 409 558 A (TAKAHASI RYOICHI ET AL) 25 April 1995 * column 1, line 13 - column 1, line 59 * * column 3, line 60 - column 6, line 37; figures 2-24 * ---	1,2,9,10	G01R33/421
X	US 5 559 435 A (HARADA AKIHIRO) 24 September 1996 * column 1, line 6 - column 1, line 42 * * column 2, line 57 - column 5, line 8; figures 1,2,5 *	1,2,7	
A	---	3,6,8,10	
A	US 5 349 297 A (DEMEESTER GORDON D ET AL) 20 September 1994 * column 4, line 28 - column 6, line 15; figures 1-3 * ---	1-10	
A	US 5 572 131 A (RZEDZIAN RICHARD) 5 November 1996 * column 4, line 1 - column 5, line 46; figures 2,4 * -----	1-4,9,10	TECHNICAL FIELDS SEARCHED (Int.Cl.6) G01R

The present search report has been drawn up for all claims

Place of search	Date of completion of the search	Examiner
MUNICH	23 November 1998	Lersch, W

CATEGORY OF CITED DOCUMENTS

X : particularly relevant if taken alone
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E : earlier patent document, but published on, or after the filing date
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